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ON HEEL PAD ELASTICITY MODELING

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INTRODUCTION

The human heel pad anatomically consists of a complex structure with neuronal, vascular, fibrous and elastic components that are intertwined with fat cells [1]. It presents anisotropic non-linear visco-elastic characteristics as with the majority of the human soft tissues, and it acts as an efficient shock absorber reducing the impact forces during gait. The biomechanical aspects of the heel pad are still under investigation particularly when there is a disorder.

Trauma to the heel pad and/or diseases of the heel may cause the “destruction” of its intricate septation which results in permanent damage of its shock absorbency capability. Specifically, *falanga torture*, a widespread corporal punishment in many countries, represents prolonged repetitive high-energy injury to the heel [2]. The effects related to this kind of torture are very difficult to identify medically, even though there is permanent damage that leads to impaired gait and chronic pain [3].

Palpation is still widely used as an important diagnostic tool. The standard medical practice of soft tissue palpation is based on qualitative assessment of the low-frequency stiffness of the tissue. Unfortunately, this qualitative technique, as well as MRI, is not powerful enough for a medico-legal assessment of *falanga torture*. In order to obtain a quantitative evaluation, and to gather information on the complete deformation of the tissues it is necessary to develop and use a device based on an indentation experiment. However, this alone does not provide information which allows a description of the tissue damage at a microscopic level. A computational simulation of the heel pad has also to be done, in order to have such description.

The aim of the present work is to develop heel pads model in order to simulate the biomechanical behavior (both in healthy and damaged cases) when subjected to a known external compression.

METHODS

The 3D finite element model of the heel pad, based on real MRI data, is created by assuming the constitutive law of the tissue and imposing the behavior found through indentation tests. The 3D model is then optimized by applying an iterative process comparing the results with those from the indentation test, and therefore adapting the tissues constitutive law. Once the 3D model is set up, the simulation of the heel pad can be first used to define test conditions (proper load and velocity at which the load must be applied) for real investigations, and then to relate the tissue damage to the stiffness measured with the indentation test.

In the present work the 3D finite element mesh of the heel pad was built on the basis of MRI scan data [Siemens Magnetom Trio (3T), Fat-suppressed 3D dual echo steady state (DESS) sequence with (0.7 mm)³ isotropic resolution, matrix 320x576x104. The echo and repetition time was 5.4ms and 16ms respectively and the excitation angle was 25 degrees. The acquisition time was 5 minutes.].

All MR images (DICOM format) were imported in SIMPLOWARE[®] (ScanIP 3.2, ⁺ScanFE 3.1.4) to create a solid model, as shown in Figure 1. The segmentation of the images was done by using proper thresholds and region growing, while filters as morphological close filter and recursive Gaussian smoothing filter were applied to merge the structures and to improve the quality of the final model. Segmented images were then exported to LS-DYNA Finite Element Software. In order to take into account all the non-linearities and visco-elasticities of the tissues, the hyperlastic parameters were chosen for the computational model [4]. All the values characterizing the modeled skin-muscle-soft tissue-bone complex were based on literature [5].

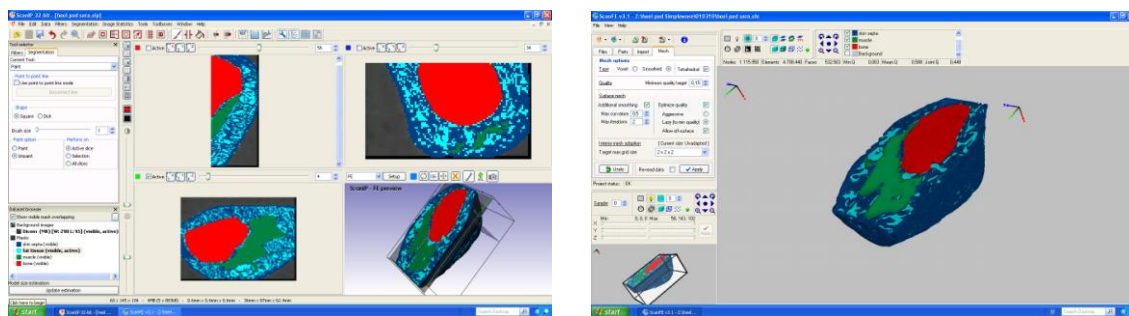


Fig. 1: 3D mesh of the heel pad created with Simpleware.

The indentation device is made of a load cell (RDP Electronics model 31) and a linear transducer (RDP Electronics LVDT Displacement Transducer), both connected to amplifiers (RDP Electronics E725) and assembled inside an aluminium cylindrical body. An additional large screw is needed for the indentation test. This is placed at the bottom of the cylindrical body. A circular shaped sensor ($\Phi=40$ mm) is attached to the load cell at the other extremity (see Figure 2). The more the screw is tightened in, the more compression is applied to the heel pad by the sensor. The real displacement of the heel pad tissue starts from when the sensor touches the skin of the heel and the load cell shows an increasing load. Until the load is zero, the recorded displacement is only the feed of the sensor. For the present experiment five compressions are applied with an interval of 10 minutes between each trial, allowing the tissue to relax. Data are then acquired and processed by using LabVIEW and Excel.



Fig. 2: Indentation device.

DISCUSSION

So far the crucial point of the heel pad modeling is the design of the intricate internal structure of the heel fat pad. Specifically, the septal walls and the fat cell chambers should be simulated with their characteristic honeycomb microstructure [1]. Such microstructure could be built by doing a computational sub-modeling that would allow to define the constitutive laws describing the behavior of the heel pad tissues at the microstructural level. By applying this procedure it will be possible to differentiate diseased tissues from healthy ones. Indeed, the mechanical integrity and functioning of the heel pad as a whole is dependent on the integrity of the septa that enclose each independent chamber [1].

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